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Assisting gait with free moments or joint moments on the swing leg

Saher Jabeen*† & Andrew Berry*, Thomas Geijtenbeek, Jaap Harlaar, and Heike Vallery

***Abstract*— Wearable actuators in lower-extremity active orthoses or prostheses have the potential to address a variety of gait disorders. However, whenever conventional joint actuators exert moments on specific limbs, they must simultaneously impose opposing reaction moments on other limbs, which may reduce the desired effects and perturb posture. Momentum exchange actuators exert free moments on individual limbs, potentially overcoming or mitigating these issues.**

We simulate unperturbed gait to compare conventional joint actuators placed on the knee or hip of the swing leg, and equivalent angular momentum exchange actuators placed on the shank or thigh. Our results indicate that, while conventional joint actuators excel at increasing toe clearance when assisting knee flexion, free moments can yield greater increases in stride length when assisting knee extension or hip flexion.

***Index Terms*— Rehabilitation robotics, Fall prevention, Leg orthoses and prostheses, Free moments, Joint moments**

I. INTRODUCTION

Falling is a major contributor to mortality and morbidity amongst older adults [1], [2]. Although numerous causes for loss of balance have been identified, impaired balance control is known to inhibit recovery from trips or slips amongst older adults [2], [3] and other vulnerable populations such as stroke survivors [4] and transfemoral amputees [5].

Visual and cognitive impairments can increase the risk of falling amongst the elderly [3]. Loss of muscle strength and slow response time degrade the ability to recover from perturbations, and are known to be related to aging [3], [6]. Balance recovery after perturbation, such as tripping over an obstacle while walking, often requires performing strategic adjustments to swing leg movement [7], such as increasing foot ground clearance or stride length.

Sufficient swing foot clearance during gait is crucial to avoid tripping or stumbling over small obstacles and, hence, falling [8], [9]. With age, the ability to effectively control the lower limbs to avoid contact with the ground or obstacles reduces significantly [9]. Of particular relevance is the height of minimal toe clearance (MTC), which occurs when the swing foot velocity is high and the projected body centre of mass (CoM) leaves its base of support. Insufficient MTC hence increases the risk of tripping while walking on uneven terrain and can lead to falls [9], [10], particularly amongst individuals with highly variable foot trajectories [11].

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Foot placement has been recognized as one of the primary means of balance control, and concepts such as the 'extrapolated centre of mass' [12] and 'foot placement estimator' [13] have been used to both explain human behaviour and quantify (in)stability of stance and gait. During gait, these models determine that proportional increases or decreases of step length are capable of countering the effects of either a forward or backward CoM perturbation. However, control of the swing leg and foot placement can be impaired, resulting in greater risks of falling and injury. Older adults often have weaker stance leg push-off and compensate by increasing hip flexion on the swing leg [14], while sufferers of osteoarthritis face difficulties extending the knee quickly enough during normal walking [15]. When perturbed, the elderly typically exhibit stepping responses similar to the young, but their reactions are often delayed and require multiple subsequent corrective steps to regain balance [16], and they have greater difficulty adjusting step length to avoid obstacles [17].

Falls might be prevented with assistance from a wearable robot. Wearable robots, including active orthoses and prostheses, have gained considerable interest for both therapy [18], [19] and mobility assistance [20]. However, greater focus has thus far been devoted to wearable aids that compensate for muscle paralysis or weakness than to those targeting milder impairments that require only subtle assistance of limb motion or balance (the latter of which is often completely neglected). Such a minimalistic aid might actively prevent loss of balance by augmenting the passive dynamics of the swing leg [21] to, e.g., guide the foot over an obstacle or ensure that the swing leg is extended far and fast enough to avoid stumbling. To limit the mass of this aid, it might even actuate during *only* the swing phase: since the swing leg has relatively little inertia in comparison to the whole body, moments applied to the swing leg may induce larger kinematic changes than those on the stance leg and, potentially, contribute greater to overall balance. We examine here how such a system might assist flexion of the knee or hip during the early swing phase to increase toe clearance, or assist knee extension or hip flexion in the mid-late swing phase to increase forward foot placement (stride length).

However, controlling balance with wearable robots remains challenging due to their limited ability to manipulate the angular momentum of the whole body, a key determinant of bipedal stability [22], [23]. To influence limb motion, wearable robots, per-definition, lack a fixed connection to the ground or an inertially-fixed structure (as in treadmill-bound robotic gait trainers) against which robotic actuators can exert forces. Instead, assistance is often achieved by placing motors at the biological joints and actuating in parallel with

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the musculature by exerting opposite reaction moments on the adjacent limbs. However, these opposite moments do not directly contribute to a net change in angular momentum and risk internally perturbing posture.

Angular momentum exchange actuators (AMEAs), such as reaction wheels and control moment gyroscopes, provide exciting new possibilities for wearable robotics. Unlike conventional actuators, which exert opposing moments between two bodies connected by a joint (joint moment, JM, Fig. 1a), AMEAs exert moments between a body and a rotating mass contained within the actuator, where the result is similar to a free moment (FM) or moment exerted against an inertially-fixed body. For a wearable device, this entails that (i) the actuator need not be placed on a joint, but at *any* location on a body segment, (ii) a net contribution to angular momentum can be made *even without contact with the ground*, and (iii) no opposite reaction moments are exerted by the actuator on adjacent body segments, reducing the risk of internal perturbation. This would enable, for example, a transfemoral prothesis containing an AMEA to provide assistance to the hip, even without a structure spanning the hip, which could benefit amputees exhibiting gait asymmetry due to muscle atrophy around the residual joint [24].

Wearable AMEAs comprising reaction wheels or control moment gyroscopes have been described in backpack-like balance aids [25]–[28], while others have envisaged them placed on the limbs for either emulation of a viscous environment [29], actuating or replicating lost function in upper extremity prostheses [30], [31], or assisting knee and hip flexion/extension [32], [33]. However, as of yet, no analysis has been conducted of how AMEAs on the lower extremities can be used to assist foot placement during recovery from perturbations, such as tripping, pushing, or unexpected change of elevation, and no comparison has been made between joint and free moments for this purpose.

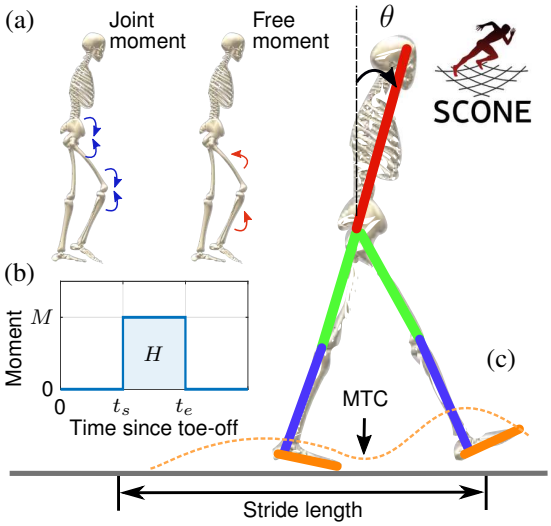

Fig. 1: (a) Joint and free moments assisting swing leg knee flexion/extension and hip flexion. (b) Applied moments parameterized as a rectangular profile of magnitude M beginning at time t_s and sustained for duration $\Delta t = t_e - t_s$, imparting angular impulse $H = M\Delta t$. (c) The 7-segment walking model and outcome measures: stride length, minimum toe clearance (MTC), and trunk pitch (θ).

We aim here to (i) determine whether free moments offer tangible performance benefits over joint moments for influencing swing-leg kinematics, and (ii) investigate the placement of such actuators to maximize any such benefit. We present a simulated comparison of joint moments applied separately to the knee and hip and corresponding free moments applied to the shank and thigh, respectively, to evaluate the extent to which step length and toe ground clearance can be increased during unperturbed gait. We hypothesize that, alongside advantages relating to freedom of actuator placement, AMEAs will be capable of performance benefits not achievable with equivalent joint actuators.

II. METHODS

Simulations were performed using a 7-segment (foot, shank, thigh, trunk; Fig. 1c) sagittal-plane musculoskeletal model. The model contains 14 Hill-type musculotendon units (iliopsoas, vasti, tibialis anterior, gluteus maximus, hamstrings, gastrocnemeus, soleus) configured according to Delp *et al.* [34] and using muscle dynamics according to Millard *et al.* [35]. Each foot segment contains two contact spheres that generate friction and restitution force [36].

To generate walking patterns, we used the reflex-based walking controller by Geyer and Herr [37] to produce muscle excitation patterns over time. Both the initial state (7 parameters describing the internal degrees of freedom) and the control parameters (29 parameters) were optimized simultaneously using Covariance Matrix Adaptation [38]. The optimization minimizes cost-of-transfer [39] at a minimum speed of $1\,\mathrm{m/s}$, and avoids knee hyper-extension through penalty forces. The controller and the optimization were implemented and performed using SCONE [40], using OpenSim [41] for the underlying dynamics simulation[1].

After the 11$^{\text{th}}$ heel strike of the left leg, either a joint moment (JM) or free moment (FM) was applied to one of two locations on the right leg, inducing primarily either knee flexion (KF), knee extension (KE), or hip flexion (HF). Knee flexion/extension could be realized with an actuator on the knee (JM) or shank (FM), and hip flexion with an actuator on the hip[2] (JM) or thigh (FM), as shown in Fig. 1a. For this preliminary analysis, the masses of the hypothetical actuators are assumed to be negligible after neurological adaptation. Changes in mass due to actuator type or capability (e.g. maximum moment) and changes in biomechanics due to

[1]Our walking controller was optimized using a specific initial guess, which was downloaded as part of the SCONE software [40]. It is important to acknowledge that different initial guesses or random seeds will produce slight variations in optimized gait patterns. Even though we believe these variations will not have a major effect on our outcomes, this could be an interesting avenue for future research.

[2]We have represented this as moments between the swing leg and trunk, but could alternatively be between the swing and stance leg. We did not investigate how this alternative would affect the dynamics.

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placement of this mass at different locations on the body are left for future investigation.

All moment profiles were parameterized as a rectangular profile beginning at time t_s after toe-off of the right leg, with magnitude M and duration Δt (Fig. 1b) – this shape was taken for convenience and is not claimed to be optimal. The parameters were discretized into 3D grids for analysis, with $M \in [5, 30]\mathrm{N\,m}$ ($5\,\mathrm{N\,m}$ increments) intended to represent realistic capabilities of a wearable actuator, and $\Delta t \in [50, 300]\mathrm{ms}$ ($50\,\mathrm{ms}$ increments) was selected to span the majority of the swing phase (approx. $530\,\mathrm{ms}$). For assisting knee flexion, moments were applied only in the early-to-mid swing phase (approx. 0-$200\,\mathrm{ms}$), so Δt was truncated to $[50, 200]\mathrm{ms}$ and the start time selected as $t_s \in [60, 180]\mathrm{ms}$ ($20\,\mathrm{ms}$ increments). Knee extension was performed in mid-to-late swing phase and hip flexion in early-to-late swing phase, so $t_s \in [180, 290]\mathrm{ms}$ ($20\,\mathrm{ms}$ increments) and $t_s \in [120, 240]\mathrm{ms}$ ($20\,\mathrm{ms}$ increments) were selected, respectively.

Changes in stride length (measured between successive heel strikes of the same foot) and minimum toe clearance (MTC) of the actuated swing leg were selected as the primary outcome measures for comparing JM and FM in each application (KF, KE, HF). For the purposes of this analysis, the parameter t_s is not of interest, so was selected to maximize either MTC (KF, HF) or stride length (KE, HF) for each actuator type and each combination of parameters M and Δt. To prevent artifacts (e.g. a null-space) in the non-optimized outcome measure, t_s was computed as:

$$t_s = \arg\max\left(\lambda\,\mathrm{MTC} + (1-\lambda)\mathrm{SL}\right)\ , \tag{1}$$

where $\lambda = 0.99$ to (primarily) maximize MTC or $\lambda = 0.01$ to maximize stride length (SL).

Quantification of differences between actuators and applications was simplified by expressing the control action in terms of a single variable: during preliminary simulations, it was noted that the primary outcome measures (particularly stride length) were strongly correlated with the angular impulse imparted by the actuator, H, computed as

$$H = \int_{t_s}^{t_e} M(t)\,\mathrm{d}t = M\Delta t\ . \tag{2}$$

This can be interpreted as a measure of 'control effort' and is a key parameter in the specification and design of AMEAs.

As a secondary outcome measure, the trunk pitch angle with respect to the vertical (θ in Fig. 1c) was computed and compared with the baseline (no moment) condition.

III. RESULTS

For assisting knee flexion, it was observed that both actuator types successfully increased minimum toe clearance (Fig. 2a, positive linear correlation with H in Table I). For the same actuation effort, the free moments resulted in $28\,\%$ smaller improvements than joint moments, and negatively impacted the stride length, which the joint moments did not. However, for assisting knee extension (Fig. 2b, Table I), free moments resulted in increases in stride length a factor of 9.3 larger than with the corresponding joint moment. Because assistance was provided mostly after mid-swing, the impact on MTC was small. For assisting hip flexion, both actuator types maximized either MTC when t_s was early (Fig. 2c, Table I) or stride length when t_s was late (Table I); for all t_s, both actuators yielded similar toe clearance, but free moments gave a $20-30\,\%$ greater increase in stride length.

For knee flexion/extension, neither actuator type appreciably affected peak trunk pitch. However, for hip flexion, in which the joint moment was exerted between the thigh and the trunk, joint moments resulted in an increase in forward trunk pitch by a maximum angle of $3.4°$, compared with $0.6°$ for the equivalent free moment.

IV. DISCUSSION

It was found that, in general, it is possible to increase toe clearance and stride length either simultaneously or independently with either joint moments or free moments. However, due to the fundamentally different actuation principles and functional requirements of each joint during different phases of the gait cycle, each has distinct advantages for specific use-cases. Hence, for brevity, the two research questions – selection of (i) an actuator type and (ii) its placement – are addressed within the context of improving primarily toe clearance *or* stride length. Ultimately, the actuators are differentiated by whether the reaction moment exerted on the adjacent limb by the joint actuator is useful or harmful.

Toe clearance reaches a minimum at mid-swing and depends on the dynamics of the early swing phase. To increase toe clearance requires shortening the distance between the toe of the swing leg and the hip, which can be accomplished to varying degrees via either ankle dorsiflexion or knee flexion – while the former may be addressed with either passive [42] or active [43] ankle-foot orthoses, we focus here on active control of the knee or hip. Assisting knee flexion with joint moments was found to increase both toe clearance and stride length, while equivalent free moments applied to the shank also increased toe clearance (to a lesser degree) but decreased stride length substantially. Since the external moments applied to the shank are the same in both cases, it is thus evident that the joint reaction moments applied to the thigh fulfill a useful function for knee flexion. Shank rotation alone causes the foot to move upwards and backwards, which, although improving toe clearance, imparts angular

TABLE I: Linear dependency of stride length and toe clearance on angular impulse*

	Δ Toe clearance		Δ Stride length	
	JM	FM	JM	FM
$\mathrm{KF_{MTC}}$	36.2 (0.994)	25.2 (0.968)	6.6 (0.636)	−115.6 (0.985)
$\mathrm{KE_{SL}}$	0.2 (0.046)	0.1 (0.046)	4.5 (0.957)	41.7 (0.993)
$\mathrm{HF_{MTC}}$	4.6 (0.792)	4.6 (0.803)	22.4 (0.929)	26.8 (0.938)
$\mathrm{HF_{SL}}$	0.9 (0.498)	0.9 (0.546)	29.5 (0.948)	38.1 (0.975)

* Slope of linear fit with R^2 in parentheses. Units of slope are $\mathrm{mm/Nms}$. Shown are knee flexion (KF), knee extension (KE), and hip flexion (HF) for both joint moments (JM) and free moments (FM). Subscripts indicate whether t_s was selected to maximize either minimum toe clearance (MTC) or stride length (SL).

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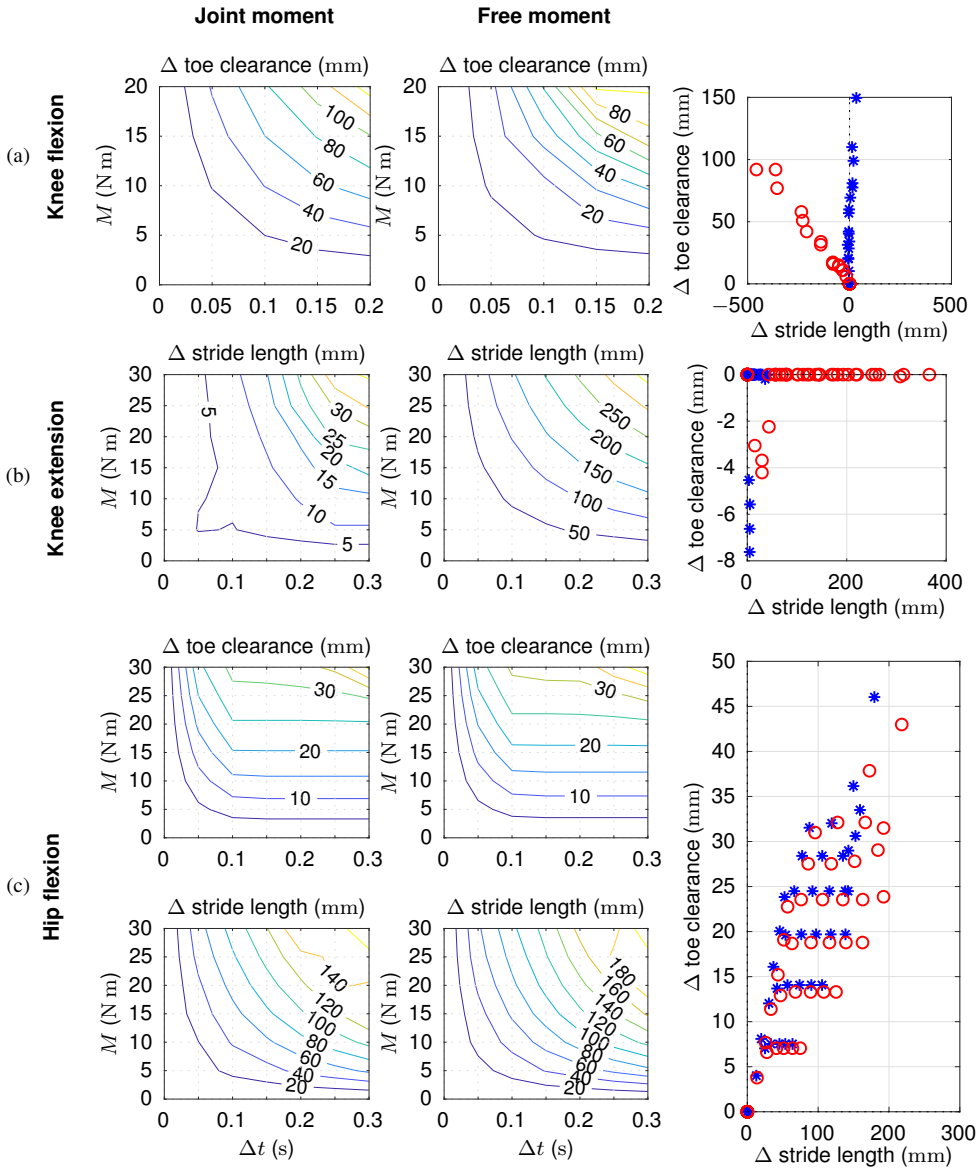

Fig. 2: (a) Knee flexion, (b) knee extension, and (c) hip flexion, where t_s is selected to maximize toe clearance, stride length, and toe clearance, respectively. Isoplots show only select quantities (see titles) for joint moments (left) and free moments (centre). Scatter plots (right) show both toe clearance and stride length for joint moments (*) and free moments (○).

momentum that inhibits knee extension in the late swing phase. Adding an opposite reaction moment to the thigh induces hip flexion, which raises the knee and moves the foot forward, benefitting both toe clearance and stride length simultaneously. To increase toe clearance without inhibiting stride length, conventional joint actuators may be preferred.

Stride length is influenced by several factors: propulsive forces generated by hip extension and plantar flexion during late stance phase, and hip flexion and knee extension at the mid-late swing phase. By assisting knee extension only, we found that free moments gave substantially greater increases in stride length than the equivalent joint moments. In contrast to knee flexion, the reaction moments exerted on the thigh by the joint moments had the tendency to induce hip extension, which slowed the forward motion of the foot and reduced the effect of the shank rotation. Hence, in this application, a joint reaction moment at the knee is undesirable and a free moment applied to the shank would be preferred.

Assisting hip flexion was found to benefit both toe clearance and stride length, and could be adjusted to increase either by selecting onset times either earlier or later in the swing phase, respectively. Hip flexion during early swing phase accelerates the knee forward and induces knee flexion, thereby increasing toe clearance, and in the late swing phase raises the thigh and induces knee extension, improving stride length. In this, the differences between the two actuators were relatively small, but greater stride length was possible with free moments. The greater similarity in this case was

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due to the fact that the reaction moment of the joint actuator was exerted against the trunk, which has relatively high inertia and is stabilized by the stance leg, so did not induce severe internal perturbation as did the knee on the thigh. Nevertheless, it was observed that this resulted in the joint moment actuator producing a forward trunk pitch that could potentially hamper balance recovery [44]. For increasing stride length, joint moments were more effective when assisting hip flexion than knee extension, but free moments performed better overall and were similarly effective at either the shank or thigh.

We have thus far neglected the actuator mass. Without accounting for its mass, a $2\,\mathrm{N\,m\,s}$ AMEA on the thigh, for example, could increase stride length by $102\,\mathrm{mm}$, or a $13\,\%$ increase over a nominal step length of $752\,\mathrm{mm}$. For this angular impulse, we estimate that a control moment gyroscope AMEA, in which approximately $50\,\%$ of its mass is concentrated in the rim of a rotor of diameter $80\,\mathrm{mm}$ rotating at $20\,\mathrm{krpm}$, could have a total mass of $1\,\mathrm{kg}$. With this mass added to both thighs and the neurological control parameters re-optimized, the nominal step length decreased to $749\,\mathrm{mm}$ and actuated increase of stride length decreased to $92\,\mathrm{mm}$, or $12\,\%$ of step length. While the consequences of neglecting mass are small in this example, we expect them to be greater for heavier actuators or those placed more distally on the body.

Finally, the choice of actuator type also has implications for the usability of the aid. Many systems targeting paralysis or muscle weakness have attempted to control (or constrain) the entire kinematic chain between the CoM and the ground, often achieved by fastening parallel 'exostructures' to the lower limbs. However, for persons with mild impairments requiring subtler assistance for movement and balance, the benefits of such systems might be overshadowed by drawbacks such as misalignment between exoskeletal and biological joints, resulting in uncomfortable or unsafe constraint forces [45], or poor usability due to the difficulty of donning and doffing devices with numerous points of attachment [46]. Recent efforts have attempted to make these structures less obtrusive by incorporating self-aligning joints [47] or minimize their placement around the joints by either reducing the actuated degrees of freedom [48], [49], replacing rigid linkages with compliant attachments to the body (so-called *soft exosuits*) [50], [51], or designing non-anthropomorphic support structures that are (partially) decoupled from the legs [52]–[54]. AMEAs may offer additional opportunities, and possess unique benefits such as (i) the freedom to place actuators away from biological joints, simplifying attachment to the body, and (ii) the ability to exert moments in any arbitrary axis, including those that vary with time or do not align with a specific joint, perhaps allowing, e.g., combined assistance to both flexion/extension and adduction/abduction of the hip.

V. CONCLUSION

We presented here the first analysis of how angular momentum exchange actuators (AMEAs) may be applied to assist the swing leg during gait. While AMEAs have the practical benefit that they do not need to be collocated with biological joints, we have shown that they additionally have the potential to yield greater control over stride length than conventional joint actuators. However, for increasing toe clearance via knee flexion, the reaction moment exerted on the thigh by a joint actuator was found to beneficially induce hip flexion and compensate for reduced knee extension in the late swing phase. Future research may investigate other strategies, such as the combination of knee flexion and extension to increase both toe clearance and stride length.

Although we have seen that key gait parameters affecting balance recovery can be influenced, we do not yet know how this can be meaningfully applied in response to a perturbation or in actual human individuals. Future work will also investigate the effectiveness of swing leg actuation for balance recovery from perturbations such as trips, pushes, and unexpected changes in elevation. In this, it is of interest to investigate how either actuator can influence the whole-body dynamics and, in particular, angular momentum, which is known to be important for stability [3], [22], [23]. In addition, we hope to address lateral balance, which requires greater active control than the sagittal plane [55], [56].

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